Wireless Power Supply for Implantable Biomedical Device Based on Primary Input Voltage Regulation

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***Abstract*-This paper presents a wireless power supply system for implantable biomedical devices. Magnitude of the input voltage supplied to the primary power converter is dynamically regulated according to the power demand of the device. The major advantage of such a system is that its average power loss is minimized. Unlike methods implemented at implantable secondary (pick-up) side, the magnitude regulation is undertaken at the external primary side. Thus the heating effect and physical size of the implantable secondary can be reduced. The system utilizes parallel tuning circuit to boost the voltage induced in the secondary pick-up, and does not require a tight coupling between the primary and secondary coils. As a result, the system has great tolerance to the variation in the air gap distance between the coils. The characteristics of the magnitude regulated power flow have been thoroughly analyzed, and both simulations and laboratory experiments have verified the proposed system.**

I. INTRODUCTION

Implantable biomedical devices have found applications in a wide range of biomedical areas, including pacemakers, monitoring devices, functional electrical stimulators (FES), left ventricular assist devices (LVAD), and artificial hearts. Supplying power to these devices for long-term operation is a challenging task. Traditionally, implantable batteries and percutaneous link power supply systems are used. However, the batteries have limited energy storage and life span, and the percutaneous links across the skin impose infection risks. Wireless systems have been developed to supply power over relatively large air gaps to implantable biomedical devices [1-5]. Compared to the devices relying purely on implantable batteries, the biomedical devices driven by the wireless power supplies can be much smaller, especially when high power outputs are required.

A major issue involved in the wireless power supply systems is power flow regulation. This is particularly important to systems with high coupling variations and load changes. In biomedical applications, it is a common practice to implement a power regulating unit at an implantable pick-up (secondary) side [6-8]. However, this requires extra components which may contribute to high heat generation and will contribute to additional size and weight. To reduce the power losses and physical size of the pick-up circuit, a closed-loop power control strategy is a favorable choice to regulate power flow at the external primary side of the system.

In this paper, a wireless power supply system based on voltage magnitude regulation at the external primary side is presented. The power flow of the system is regulated by controlling the input voltage magnitude of the primary power converter according to power demands of load and the dynamic changes of circuit parameters.

II. PROPOSED SYSTEM

Fig. 1 shows the basic configuration of the magnitude regulated wireless power supply system. At the external primary side, a current-fed push-pull resonant converter is employed due to its advantages of high efficiency, low cost and small physical size. The operating frequency of the system is governed by the zero voltage switching frequency of the push-pull resonant converter, which is determined by [9]:

$$\omega = \left(\sqrt{\left(1 - \frac{1}{4Q^2}\right) \Big/ \left(L_p C\right)} - \frac{\varphi}{T} \right) \Big/ \left(1 + \frac{2\varphi}{\pi}\right) \tag{1}$$

where φ, Q and T are the initial phase angle, the quality factor, and the time constant of the converter.

Inductor L_p represents primary coil (track), which is tuned with a capacitor C, forming the primary resonant tank. The pick-up (secondary) coil represented by L_S is implanted for inducing power from the electromagnetic field generated by the primary coil. Since L_s is tuned with capacitor C_t in parallel (C_t=1/($L_s\omega^2$)), the induced voltage in the pick-up coil can boost up according to the boost-up factor of the pick-up [10]. Due to the boost up ability, air core windings can be utilized in the primary and secondary coil while a relatively large gap distance between them is allowed. Additional weight and heating effect caused by magnetic cores are eliminated in such a air-core system which is with parallel tuning.

As shown in Fig. 1, for achieving the maximum power transfer capacity, a full bridge rectifier is adopted in the pick-up to convert ac power to dc. Also, a dc inductor L_{dc} is employed at the dc side of the rectifier for increasing power delivery ability. The dc inductor maintains the current flow through the rectifier to be continuous so that the power transferred from ac to dc side becomes more stable [10].

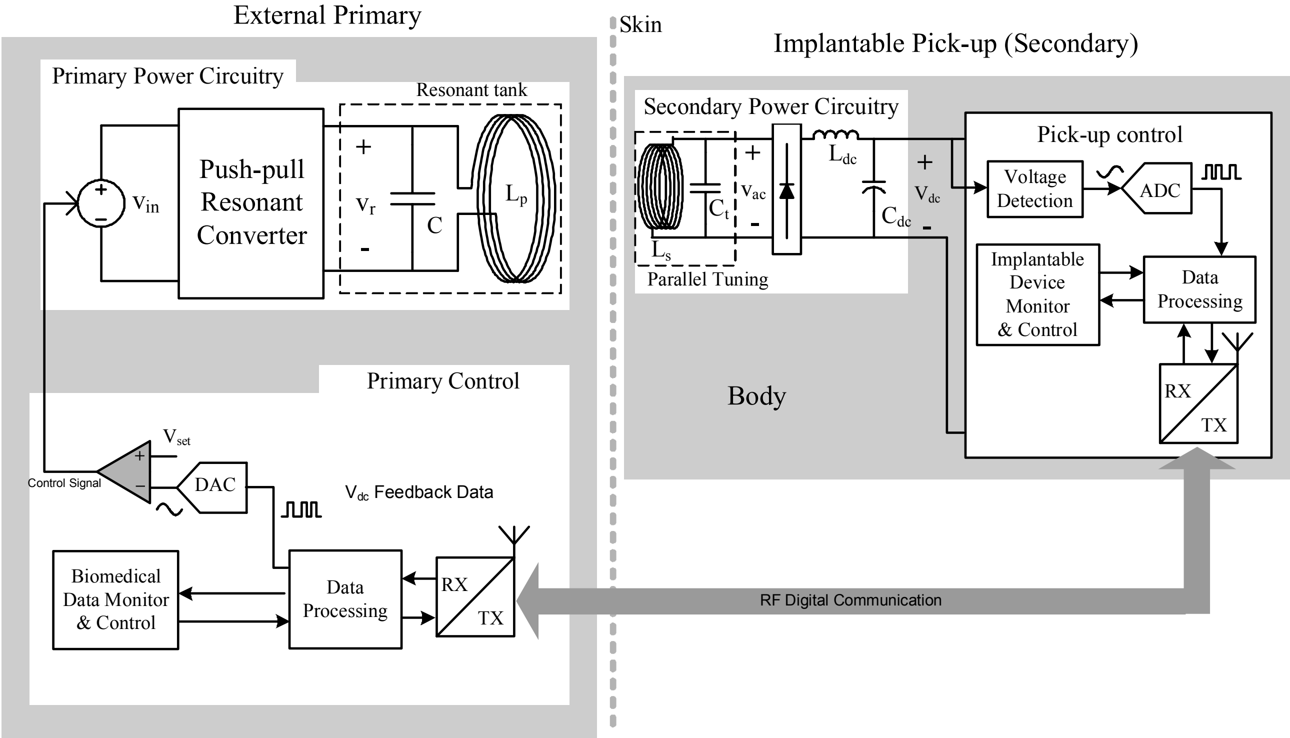

Fig. 1: Wireless power supply system with magnitude control from external primary side.

The primary current-fed push-pull converter is powered by a variable dc voltage source, v_{in}. The power flow in the system can be controlled by varying the value of v_{in} according to the power demand of biomedical devices. This variable dc voltage supply can be achieved by applying a dc-dc power converter such as a buck or buck-boost converter.

In addition to transferring physiology signals, the RF data transceivers are also used to transfer the signal of the output voltage V_{dc} from the implantable secondary to external primary, as shown in Fig. 1. The value of the input voltage v_{in} varies depending on the feedback signal of V_{dc} to regulate the power flow so as to maintain the voltage supplied to the implantable device constant. Due to digital communication, the output voltage signal has to be digitized using analog to digit converter (ADC) before it is transmitted from the implantable secondary. On the contrary, after the receiver in the primary receives the digital feedback signal of V_{dc}, a digit to an analog converter (DAC) converts it back to analog signal. Then it is compared to the preset nominal voltage V_{set} to generate a control signal for controlling the value of the dc voltage v_{in}.

III. ANALYSIS OF MAGNITUDE REGULATED POWER FLOW

It has been known that the magnitude of the resonant voltage (v_r) across the primary coil is proportional to the magnitude of the input dc voltage (v_{in}) of the push-pull converter according to [9]:

$$\hat{V}_r = \pi v_{in} \tag{2}$$

where $\hat{V}_r$ is the peak value of the resonant voltage v_r.

And, when the pick-up operates under the maximum power transfer condition, the current flow through the rectifier would be continuous. In this situation, the secondary resonant voltage (v_{ac}) across the pick-up coil is determined by the output dc voltage (V_{dc}) of the secondary side according to [10]:

$$\hat{V}_{ac} = \frac{\pi}{2} V_{dc} \tag{3}$$

where $\hat{V}_{ac}$ is the peak value of the secondary resonant voltage (v_{ac}) across the pick-up coil L_s.

Equation (2) and (3) show that in the maximum power transfer situation the amplitudes of the resonant voltages across the primary and secondary coil are independently determined by the input dc voltage of the primary converter and the output dc voltage of the secondary pick-up. Therefore, the maximum power capacity can be analyzed using the model shown in Fig. 2.

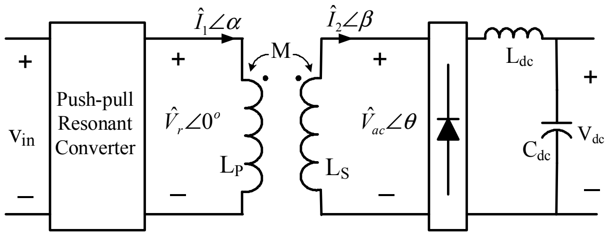

Fig. 2: Model of coupling between primary and secondary.

Inductors L_p and L_s represent the primary and secondary coils. The voltages across them are the primary and secondary resonant voltage, v_r and v_{ac}. L_p and L_s are magnetically coupled with a mutual inductance M. Ignoring practical high frequency harmonics and system power loss, the steady state analysis can be undertaken using phasor approach. Assuming the phase angle of the primary resonant voltage is zero, the voltage phasors across inductor L_p and L_s are $\hat{V}_r\angle 0^o$ and $\hat{V}_{ac}\angle\theta$ respectively, as shown in Fig. 2. The phasors of the current flows through L_p and L_s are defined as $\hat{I}_1\angle\alpha$ and $\hat{I}_2\angle\beta$ respectively.

Based on the model shown in Fig. 2, the following equations can be obtained:

$$\hat{V}_r\angle 0^o = (j\omega L_P)\cdot\hat{I}_1\angle\alpha - (j\omega M)\cdot\hat{I}_2\angle\beta \tag{4}$$

$$\hat{V}_{ac}\angle\theta = (j\omega M)\cdot\hat{I}_1\angle\alpha - (j\omega L_S)\cdot\hat{I}_2\angle\beta \tag{5}$$

Under the steady state condition, the open circuit voltage (v_{oc}) of the pick-up coil L_s can be determined as:

$$\dot{V}_{oc} = j\omega M\dot{I}_1 = \omega M\hat{I}_1\angle(\alpha + 90^o) \tag{6}$$

It has been known that the open circuit voltage v_{oc} and the current flowing through the pick-up coil, i_2, are in phase when the pick-up operates under the maximum power transfer condition [10]. Thus, the following relationship between phase angle α and β can be obtained:

$$\alpha + 90^o = \beta \tag{7}$$

Solving equation (4), (5) and (7), the peak value of the primary current can be expressed as:

$$\hat{I}_1 = \frac{1}{\omega}\sqrt{\frac{(L_S\hat{V}_r)^2 - (M\hat{V}_{ac})^2}{(L_P L_S)^2 - M^4}} \tag{8}$$

If the system is loosely coupled so that the mutual inductance M can be considered to be very small, (8) can be approximately expressed as $I_1 \approx V_r/(\omega L_P)$, where I_1 and V_r are rms values of current i_1 and voltage v_r. This means that the primary current is equal to the primary resonant voltage divided by impedance of the self-inductance of the primary coil.

Then, the peak value of the open circuit voltage v_{oc} of the pick-up coil can be expressed as:

$$\hat{V}_{oc} = \omega M\hat{I}_1 = M\sqrt{\frac{(L_S\hat{V}_r)^2 - (M\hat{V}_{ac})^2}{(L_P L_S)^2 - M^4}} \tag{9}$$

The maximum power transferred by the pick-up is determined by the following equation [10]:

$$P_{\max} = \frac{\pi}{2\sqrt{2}} I_{sc} V_{dc} \tag{10}$$

where V_{dc} is the output dc voltage of the pick-up, and I_{sc} is short circuit current of the pick-up coil ($I_{sc}=V_{oc}/(\omega L_s)$).

According to (9) and (10), the maximum power transfer capability of the system can be obtained as:

$$P_{\max} = \frac{\pi M V_{dc}}{16 L_S f}\sqrt{\frac{(2L_S v_{in})^2 - (MV_{dc})^2}{(L_P L_S)^2 - M^4}} \tag{11}$$

Considering that the coupling coefficient k ($k^2=M^2/(L_pL_s)$) of the wireless power supply system is only up to 0.1, the value of $(L_pL_s)^2-M^4$ is approximately equal to $(L_pL_s)^2$. Then, the maximum power of the system determined by (11) can be calculated using the following equation:

$$P_{\max} = \frac{\pi M V_{dc}}{16 f L_P L_S{}^2}\sqrt{(2L_S v_{in})^2 - (MV_{dc})^2} \tag{12}$$

It can be seen from (12) to obtain a meaningful solution of P_{max}, the term $2L_sV_{in}$ must be larger than MV_{dc} ,meaning equation (12) is valid to analyze the maximum power transfer capacity only when the following condition is met:

$$v_{in} > \frac{MV_{dc}}{2L_s} \tag{13}$$

Fig. 3 illustrates the relationship between the input dc voltage (v_{in}) and the maximum power transfer capability (P_{max}) of the wireless power supply system. As a comparison, two different mutual inductances, M and M', are shown, where M>M'. It can be seen that the maximum power transfer capability rises with the increase of the input dc voltage. And at the same input dc voltage, the system with higher mutual inductance has higher power transfer capability.

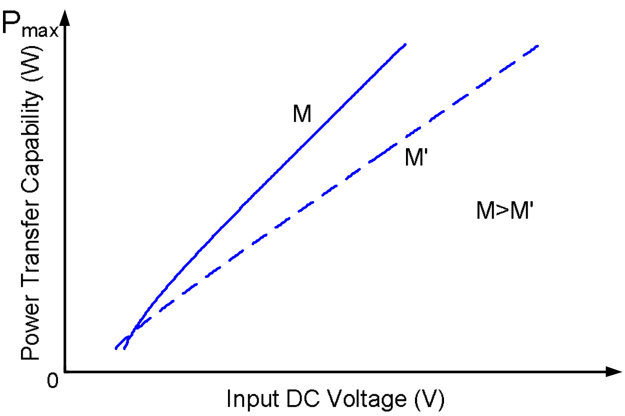

Fig. 3: Relation between maximum power transfer capability and primary input dc voltage.

It should be noted that when the dc input voltage is very low, even then the condition shown in (13) is may be satisfied, the current flowing through the pickup rectifier may become discontinuous, causing larger errors in the analysis. However, in most practical situations, the system does not operate at a very low dc voltage, so (12) is valuable for predicting the maximum possible power that the system can deliver. Another point is that the power transfer capability is not proportional to the input voltage v_{in} according to (11) and (12). However, when the input voltage v_{in} is increased to a higher value, the increase of P_{max} becomes approximately linear. This is because when v_{in} is considerably high (or M is considerably low), $2L_sV_{in}$ is much larger than MV_{dc}. As a result, Eqn (12) can be simplified as:

$$P_{\max} = \frac{\pi M V_{dc}}{8 f L_P L_S} v_{in} \qquad \textbf{(14)}$$

where power P_{max} linearly increases with v_{in}.

Fig. 3 also shows the system with a higher mutual inductance requires a smaller variation of v_{in} to regulate the power flow. On the other hand, this means that the power variation of a closely coupled system is much more sensitive to the change of the input dc voltage than a loosely coupled system.

According to (14) the design parameters such as the operating frequency f and mutual inductance M, can also affect the power transfer capability. During an actual operation, the operating frequency f and the mutual inductance M can vary due to the factors such as the air gap change caused by the movement of an animal or patient. Therefore, to maintain a constant output dc voltage from the pick-up, regulating the input voltage of the converter can also compensate for the dynamic change of the operating frequency f and the mutual inductance M.

It should be noted that although (14) shows that the maximum power transfer capability would be very high if the frequency is very low, in practice the operating frequency has to be high to reduce the physical size of reactive components such as coils and tuning capacitors. Limiting the lowest operating frequency is also for achieving a high impedance of the primary coil, so as to reduce the track current and associated power loss and heat generation. Moreover, the minimum operating frequency is restricted by the condition of the ZVS (zero voltage switching) operation of a push-pull resonant converter. This condition requires the quality factor Q of the primary coil is greater than 1.86 [9]. A too low frequency would make Q too small.

III. SIMULATION AND EXPERIMENTAL RESULTS

PLECS simulations have been conducted to verify the analysis of the magnitude control of the wireless power supply system. The main simulation parameters are shown in Table 1.

Fig. 4 shows the simulation results of the relationship between the input dc voltage of the primary converter and the system maximum power transferred. As a comparison, MATLAB calculation results according to (11) is also plotted and shown in the figure. It can be seen that the simulation results are in a good agreement with the analytical results. Since the system is loosely coupled (coupling coefficient k=0.1), the maximum output power almost linearly increases with the input dc voltage.

TABLE I
SIMULATION PARAMETERS

Parameters	Values	Notes
V_{dc}	10 VDC	Output dc voltage of secondary pick-up
f	155 kHz	Operating frequency
L_P	25 uH	Self-inductance of primary coil
L_S	4 uH	Self-Inductance of secondary pick-up coil
M	1 uH	Mutual inductance
L_{dc}	0.5 mH	Secondary DC inductance
V_{in_min}	0 VDC	Minimum input voltage of the primary
V_{in_max}	50 VDC	Maximum input voltage of the primary

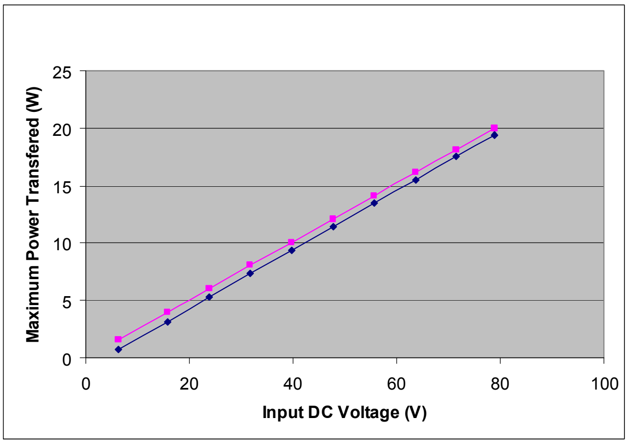

Fig. 4: Simulation results of the relationship between input voltage and maximum power transfer capability.

A prototype wireless power supply system for free moving animal implantations has also been built and tested in laboratory for verifying the system functionality. The primary coil was a lumped loop around an animal cage sized 55×30×20cm^3 (length×width×high). The electromagnetic field generated by the primary coil covers the whole cage area, but the magnetic field at the center is weaker because the distance to primary coil winding is larger. Nordic nRF24E1 transceivers were used to implement the wireless communication between the primary and secondary circuits.

Fig. 5 shows the measured results of the input dc voltage supplied to the primary converter and the output dc voltage of the secondary pick-up. By detecting the actual output voltage of the secondary pick-up, the closed loop voltage regulation system can automatically adjust the input dc voltage of the primary power converter from minimum 10 V to maximum 35 V. This will increase the track current and the flux density to a required level so that the output voltage of the secondary pick-up can be maintained constant (eg. at 7V shown in Fig. 5) across the whole animal cage. This will ensure continuous power supply to biomedical devices regardless of their actual location in the cage.

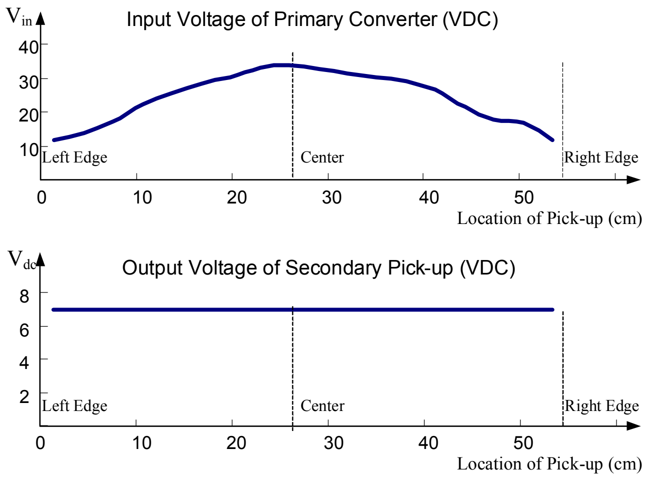

Fig. 5: Experimental results of input voltage of converter and output voltage of pick-up.

IV. CONCLUSIONS

A wireless power supply system for implantable biomedical devices based on primary voltage regulation has been proposed in this paper. The power flow of the system is regulated effectively by dynamically controlling the input voltage of the primary converter. The output dc voltage of the power pick-up is maintained constant regardless of the variations in operating frequency, loading condition and coupling coefficient, etc. The effect of the input voltage of the converter on the maximum power transfer capability has been analyzed. The analytical results show that the maximum power transfer capability is approximately proportional to the input voltage of the converter if the coupling coefficient is less than 0.1. The theoretical analysis enables the minimum value of the input voltage to be determined for achieving the maximum power transfer capacity. The simulation results are in good agreement with the theoretical analysis, and a prototype has been built to verify the operation of the system. It has shown that the proposed system can maintain the output voltage of the secondary pick-up to be constant across the full ranges of an animal cage.

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